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Active–passive biodynamics of the human trunk when seated on a wobble chair

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ABSTRACT

Unstable sitting on a wobble chair with different balance difficulty levels can be used as an effective tool in exercises as well as evaluation and therapeutic stages of rehabilitation. No data on muscle activity levels and spinal loads are however available to assess its safety compared to other regular daily activities. The goal of this study was to estimate muscle forces and spinal loads in a seated unstable wobble chair task. In vivo 3D kinematics of the trunk and seat collected in an earlier study were used here to drive computational trunk musculoskeletal models of 6 normal and 6 low-back pain subject groups sitting on a wobble chair for a duration of 10 s. Results revealed no significant differences between kinematics, muscle forces, spinal loads and force plate reaction forces when comparing these two groups. The estimated muscle forces and spinal loads were moderate though larger than those in a stationary sitting posture. Local spinal forces at the L5-S1 disc varied with time and reached their peaks (1473 N and 1720 N in compression, 691 N and 687 N in posterior–anterior shear and 153 N and 208 N in right–left shear, respectively for healthy and CLBP groups) being much greater relative to those in the stationary sitting posture (means of 12 subjects: 768 N, 284 N and 0 N, respectively). The wobble chair with characteristics considered in this study is found hence safe enough as a therapeutic exercise for both healthy and low-back pain subjects.

1. Introduction

Human body is subject to external perturbations during falls, tripping and sudden loading–unloading (Shahvarpour et al., 2014) as well as internal perturbations due to respiration (Hodges et al., 2002) and neuromuscular noise (Reeves et al., 2013). As a result, demands for muscles' passive, active and reflexive contributions increase in order to both satisfy the deteriorated transient equilibrium conditions and to maintain a sufficient margin of stability and balance to prevent falls and injuries (Panjabi, 1992). Unstable support environments such as those in standing and sitting on pivoted boards are helpful to assess and improve neuromuscular responses. Wobble chairs have been employed as a tool to investigate the trunk neuromuscular mechanisms involved in balance of the upper body in isolation from the confounding effects of the lower extremities (Cholewicki et al., 2000). Trunk stability (Freddolini et al., 2014a; Tanaka et al., 2009, 2010), trunk stiffness (Freddolini et al., 2014c; Reeves et al., 2006), neuromuscular activity (Freddolini et al., 2014b; Reeves et al., 2006), reflexive response (Radebold et al., 2001; Reeves et al., 2009) and trunk motor behavioral differences between LBP patients and healthy subjects (Willigenburg et al., 2013) have been studied using such method of unbalanced sitting.

Previous iterative kinematics-driven computational trunk model studies under sudden dynamic loads and motions with high acceleration content estimated relatively high spinal compression forces (in the range of 3–6 kN) (Bazrgari et al., 2009; Shahvarpour et al., 2015) and hence risk of low-back injuries. Initial flexed posture and antagonistic coactivity along with higher sudden load markedly increased spinal loads. Any dysfunction in the neuromuscular system associated with for example longer latency and/or muscle injury could further increase spinal loads and motions causing additional injuries. Furthermore, high spinal loads may aggravate pain in CLBP subjects that makes them excessively cautious due to the fear of pain when performing tasks (Greene et al., 2001; Khalil et al., 1987).

The primary aim of this study was to assess the safety of the wobble chair task. Despite the growing interest in unstable devices such as wobble chairs in exercises and rehabilitation therapies, no realistic model study of the trunk muscle forces and spinal loads has been carried out so far. Based on earlier in vivo measurements in

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which 18 healthy controls and 18 chronic low-back pain (CLBP) patients participated (Larivière et al., 2013), we simulated the trunk response of 12 subjects (6 controls and 6 patients) under the associated personalized trunk masses and measured kinematics. Despite the fact that no significant differences in most recorded measures (range of motion, velocity, median frequency, etc.) were found in the in vivo study between healthy controls and patients (Larivière et al., 2015), a secondary aim (exploratory study) was to compare the biomechanical measures (angular velocities/accelerations, muscle forces, spinal loads) calculated by our kinematics-driven model in an attempt to discriminate between these two groups.

2. Method

2.1. Subjects and measurements

Among 36 individuals volunteered for the in vivo study reported elsewhere (Larivière et al., 2013), 6 healthy and 6 CLBP male subjects with body height close to that in our FE model (vertical distance from the S1 to the C7 of 46.76 cm) were chosen (Table 1). A brief description of the in vivo study is provided here, with emphasis on elements specifically related to the current computational work. The inclusion criteria for CLBP subjects were: lumbar or lumbosacral pain with or without proximal radicular pain (limited distally at the knees) and presence of chronic pain defined as a daily or almost daily pain for at least 3 months. The exclusion criteria for the healthy controls were back pain in the previous year or back pain lasting longer than a week during the preceding years.

Table 1
The anthropometric data of 12 male subjects considered in this model study (range denotes the difference between max and min of data).

Subjects	Body mass (Kg) mean (range)	Body height (cm) mean (range)
Healthy controls	81.2 (29)	178 (9)
CLBP patients	82.8 (18.9)	179 (7)
All	82.0 (29)	178.5 (9)

The subjects sat on the wobble chair with the feet on an adjustable platform attached to the chair and the arms crossed on the chest (Fig. 1). The subjects were instructed to sit relaxed with the head and chest in the upright position. To avoid excessive inter subject–chair movements, feet were strapped to the chair (footstep) and thighs were secured laterally with foam cubes attached with velcro. A ball and socket pivot supported the seat, allowing for a maximum tilt of 13° in forward–backward and lateral directions (maximum range of motion allowed: 26°). The apparatus design restricted the axial rotation. Four springs (height = 4.5 cm, axial stiffness = 8467 N/m) with equal distances from the pivot were placed under the seat in front, back, right and left sides. The springs were just in contact with the seat at the beginning and did not stretch during tests as they were not attached to the seat. Kinematics of the wobble chair and trunk was measured using an Optotrak system (Northern Digital Inc., Waterloo, Ontario, Canada) at a sampling frequency of 50 Hz. Rigid marker clusters composed of three infrared light emitting diodes were attached on the seat surface and the trunk of subjects at the S1, T12, C7 and head levels. A force plate placed (AMTI, model BP900900, Watertown, MA, USA) under the chair recorded the force and the center of pressure (CoP) at 1000-Hz sampling frequency (Fig. 1).

Segment coordinate systems were defined at the S1, T12, C7 and head as rigidly attached to each segment with their orientations in the initial seated posture developed based on the global fixed coordinate system. Using each marker cluster, another segmental coordinate system was also defined. Since the orientation of both segment and cluster coordinate systems at seated posture is known in the global coordinate system, their relative rotations to the global system could be evaluated. The instantaneous orientations of the segment coordinate systems yielded segmental rotations at each instance of time with respect to the seated posture.

A simple calibration protocol (Larivière et al., 2013; Slota et al., 2008) allowed for the positioning of the springs so as to reduce the influence of body size on recorded performance. The resulting spring positions defined hence a reference system that was considered neutrally stable for each specific subject. The task difficulty was subsequently determined by adjusting the spring positions relative to those set in the foregoing reference condition. In the current study the task difficulty was set at 60% (Larivière et al., 2013). The subject was instructed to keep the eyes closed during the task. Tests started after removing the stabilizing cushions placed under the chair, but due to technical limitations, recording started ~5 s after and lasted for 60 s.

2.2. FE model studies

The three-dimensional finite element (FE) model of the spine consisted of 7 rigid bodies representing sacrum, L1–L5 lumbar vertebrae and thorax–head–arms (Fig. 1) (Bazrgari et al., 2008b; Shahvarpour et al., 2015). Based on mesh refinement

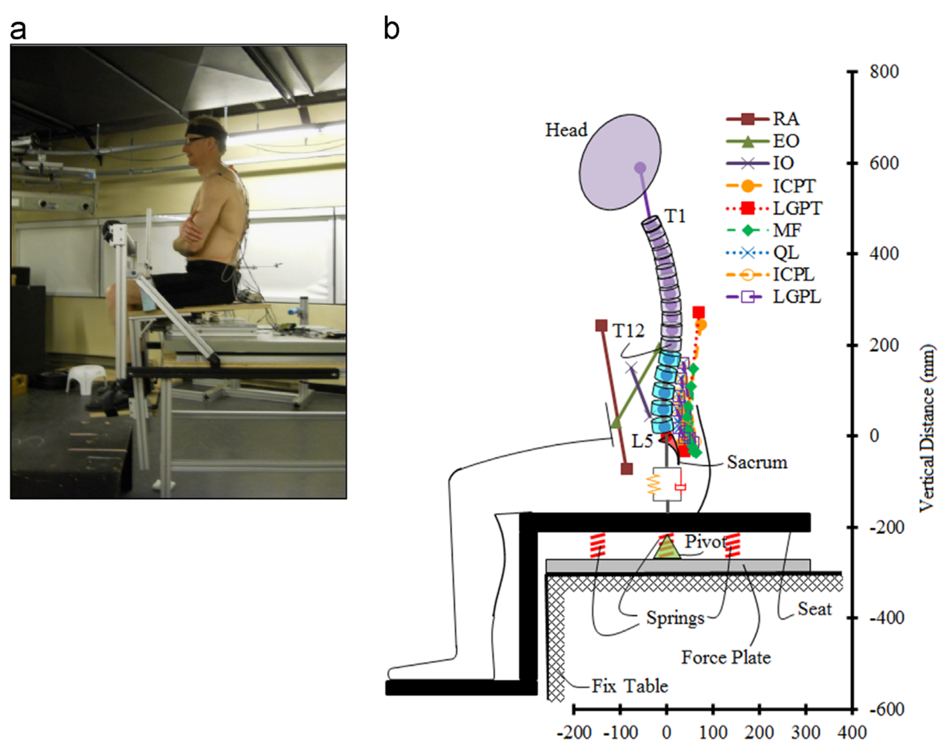

Fig. 1. The side view photo of a subject sitting on the wobble chair (A) and a schematic sagittal view of the finite element model of the subject seated on the wobble chair. RA: rectus abdominus, EO: external oblique, IO: internal oblique, ICPT: iliocostalis lumborum pars thoracic, LGPT: longissimus thoracis pars thoracic, MF: multifidus, QL: quadratus lumborum, ICPL: iliocostalis lumborum pars lumborum, LGPL: longissimus thoracis pars lumborum.